Creating 3D boundary element models of the heart from 2D projections

H. Veisterä, J. Lötjönen, and T. Katila

Laboratory of Biomedical Engineering, Helsinki University of Technology, FIN-02015 HUT, Finland

1 Introduction

Individual 3D boundary element models can be used in solving inverse problems in electro- and magnetogardiographic measurements. Many other applications exist for individual 3D models in the medical field. In some cases, MR or CT images are not available or the cost of obtaining them is too high. Modalities producing 2D data, such as X-ray projections and ultrasound images, are more readily available. Therefore, it would be useful to be able to use these images for creating 3D models.

One or two X-ray projections of an object are usually not enough to reconstruct a 3D object. Several X-ray projections of the same object from different angles are needed to create a good 3D approximation of the object. In CT, hundreds of projections are used. If only a few X-ray projections are used, some *a priori* knowledge can be incorporated to improve the reconstruction accuracy. For human anatomy, the prior knowledge, i.e. a representation of typical human anatomy, can be extracted, for example, from MR volumes. This work is based on a method where a 3D model was reconstructed from two 2D silhouettes using a prior 3D model [1].

The aim of this work was to develop an easy to use software package for creating a 3D heart model from two orthogonal X-ray projections. In addition, 3D model coordinates can be displayed on the original X-ray projections in 2D for persons who are used to locating objects from X-ray projections.

The developed application is not limited to creating only heart models but it can be applied more generally. Once a prior model of an object and two images of a new slightly different instance of the object are available, this application can be used to reconstruct the 3D geometry of the object.

2 Methods

The geometric prior model used in this work was built from magnetic resonance (MR) images. The data volume used here originates from a cardiac cine MR sequence at diastolic phase. The geometric model is built by first segmenting [2] and then triangulating [3] the cine MR volume. The segmentation is based on free-form deformation of a geometric and topologic prior model. The model is matched to the edges present in the data volume. After segmentation, a set of points is selected from the segmented volume and the points are linked with each other in such a way that triangles are produced. The Voronoi-Delaunay duality is used in the triangulation. The epicardium and both ventricles are included in the model.

The source images are sequences of X-ray images of the heart of the patient, taken at two different orthogonal angles. The images used in this work are taken during catheter operations and contain, in addition to the anatomical features, the electrodes and catheters used during the operation. The images are converted from analog video data into digital format with an image grabbing system [4].

In order to improve the low quality of the source images, a number of image-processing algorithms were implemented. The source image frames are combined with averaging and interleaving and a specifically modified local contrast enhancement algorithm is applied. The improvement in image quality helps the segmentation and visual inspection of the image. An example of the image enhancement is displayed in [Figure 1].

Several methods have been proposed for automatic segmentation of X-rays, such as active contour models [5], united snakes [6]. The quality and type of the source images, used in this work, makes them very difficult for automatic segmentation. The problem is further complicated by the fact that the object is often seen only partially in the images. A method, introduced in [1], tries to overcome these problems using prior model information and elastic matching. However, this method has not yet been tried for the type of source images used in this work. The final segmentation of the heart outline is done manually by looking at the pre-processed source image.

The outline is compared with virtual X-ray projections created from a prior 3D model, carefully segmented from MR images of the heart. The 2D vectors generated by matching the actual outlines with the prior model outlines are back-projected onto the 3D model surface. Vectors are interpolated for each model surface point. The original model points are moved according to the displacement vectors. The ventricles of the heart, not visible in the X-ray projections used, are deformed using local scaled averages of the deformations calculated for the epicardium [Figure 2].

As a result of the previous stages, necessary information exists to map 3D coordinates back to the original 2D projection coordinates. This information can be used to display 3D points on the original 2D projections, for example, on the ventricle surfaces, such as Josephson points. The points can be color-coded to display additional information.

3 Results

The quality of the digitized X-ray projections was noticeably improved and thus the heart outline segmentation was facilitated [Figure 1]. At this point the relevant features of the image are considerably more prominent than in the original X-ray image.

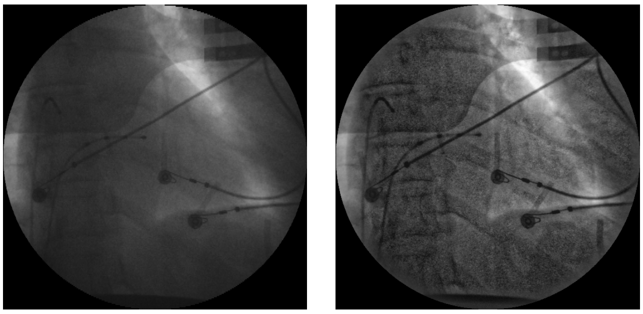

(a) (b)

Figure 1: *(a) Original digitized fluoroscopy image; (b) Image processed with frame interleaving, image averaging and local contrast enhancement*

The direct validation of the results with actual cases where both 2D silhouette images of the heart and otherwise generated 3D images of the heart could not be done since no such data was available. However, the results can be indirectly observed with just 3D data available. Silhouette images can be generated from the 3D data in various ways, for example, with virtual X-ray projections. These 2D images can then be used to generate 3D models with this application. These deformed models can then be compared with the original models. Such a comparison has been carried out for torso models in [1], where the mean error for the torso models generated was found to be approximately 1.2 voxels corresponding to about 5 mm. However, the geometric accuracy of the model is not very critical in this work, since the deformed 3D model is inspected only visually.

The accuracy of the results could be defined using similar simulations to the paper [1], as mentioned above. Because no information is available from ventricles in X-ray images, the interpretation of the ventricle surfaces is, however, only qualitative. Therefore, the geometric accuracy of the ventricles is not emphasized and studied in this work. Nevertheless, it is worth noting that since the prior model is used, the geometric relations between the epicardium and ventricles remain approximately correct during the deformation [Figure 3].

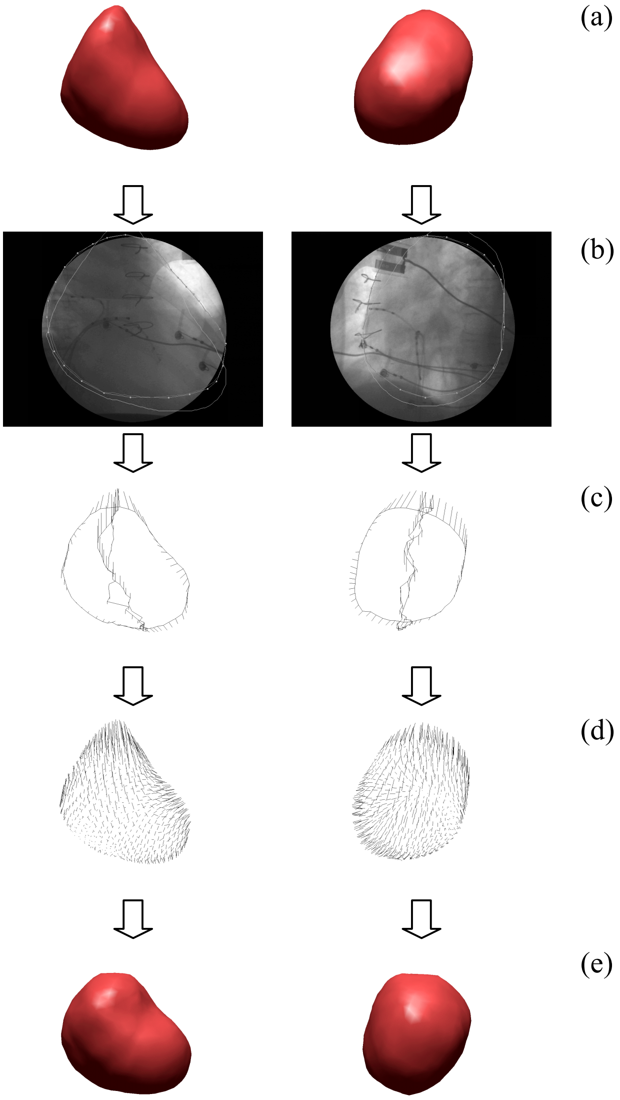


Figure 2: *The deforming process: (a) The original heart model displayed from two orthogonal angles; (b) The patient X-rays where the new heart outline is segmented from. These outlines are compared with virtual X-rays of the original model; (c) The 2D vectors generated from elastic matching between the actual outlines and the model outlines are moved onto the 3D model surface; (d) Vectors are interpolated for each model surface point; (e) The new deformed model.*

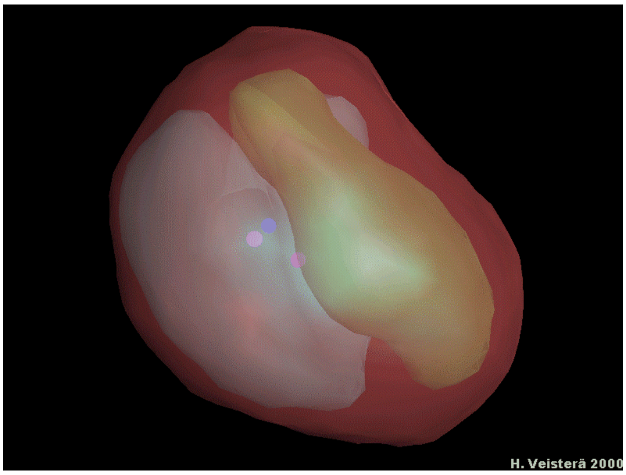


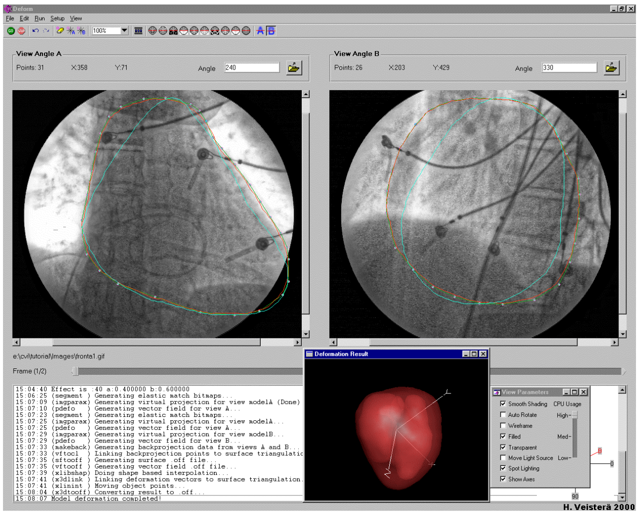
Figure 3: *A heart model created with the software including the heart ventricles.*

The information on the source imaging angle is also manually entered by the user. Changes in the resulting output when the source angles are erroneously entered are shown in Table 1. A model was created using accurate orthogonal angles and it was set as the base for comparisons. The source angles for one or both views were modified and the resulting model was compared to the base model. The difference in entered and real angles and the difference between the result and correct object volume is displayed in the table along with the overlap for the result and correct volumes. When the patient heart corresponds well to the used original heart model, the method is very robust to errors in determining the source image angles. Since the goal of the model creation is to reconstruct the correct model shape and not the size, the actual simulation parameters need not closely correspond to the actual X-ray imaging parameters used in the source X-ray projections.

Table 1: *Changes in output with different source imaging angles*

Angle diff	Volume diff.	Volume overlap
1°	+0.043 %	99.2 %
2°	+0,061 %	99.2 %
10°	-0,034 %	98.2 %
20°	+0,040 %	97.8 %

The deformation method implemented is robust and provides good results even when the source parameters contain errors. The application is easy to use and new models can be created on-line with a common desktop PC. Currently the individual anatomic heart models are used to visualize the results from a Body Surface Potential Mapping system. However, the software is not limited to creating just heart models [Figure 4].



View Angle A
Points: 31 X:358 Y:71 Angle 240

View Angle B
Points: 26 X:203 Y:429 Angle 330

e:\cvi\tutorial\Images\fronta1.gif

Frame (1/2)

```
15:04:40 Effect is :40 a:0.400000 b:0.600000
15:06:25 (segment ) Generating elastic match bitmaps...
15:07:09 (imgparax) Generating virtual projection for view modelA (Done)
15:07:10 (pdefo    ) Generating vector field for view A...
15:07:23 (segment ) Generating elastic match bitmaps...
15:07:25 (imgparax) Generating virtual projection for view modelA...
15:07:25 (pdefo    ) Generating vector field for view A...
15:07:29 (imgparax) Generating virtual projection for view modelB...
15:07:29 (pdefo    ) Generating vector field for view B...
15:07:33 (makeback) Generating backprojection data from views A and B...
15:07:33 (vftocl   ) Linking backprojection points to surface triangulatic
15:07:35 (sftooff ) Generating surface .off file...
15:07:35 (vftooff ) Generating vector field .off file...
15:07:39 (xlibshap) Doing shape based interpolation...
15:07:41 (x3dlink ) Linking deformation vectors to surface triangulation.
15:07:41 (xlinint ) Moving object points...
15:08:04 (x3dtooff) Converting result to .off...
15:08:07 Model deformation completed!
```

☑ Smooth Shading
☐ Auto Rotate
☐ Wireframe
☑ Filled
☑ Transparent
☐ Move Light Source
☑ Spot Lighting
☑ Show Axes

Figure 4: *The application user interface: the screen is divided from top to bottom into menus where options can be selected, a toolbar for quick access to frequently used options, source image views A and B for segmentation, animation controls and a message box for displaying information about the deformation process.*

4 Discussion

Presently, only one object outline can be displayed and edited at one time. If multiple object outlines could be edited at the same time, this would simplify and speed up handling of multiple objects. Also, the relations between the objects could be more easily viewed and edited.

Manual segmentation works well in practice and it can be accomplished in a few minutes. Automatic segmentation such as outlined in [1] could be implemented and tested. However, the visibility of the edges of interest in our images is much poorer than in the thorax-images used in [1]. Moreover, strong edges produced by catheters may distort automatic segmentation.

Only X-ray images were used as source images but other imaging modalities such as ultra sound imaging where the object silhouette can be seen could be used. Any 2D image where the object silhouette can be observed could be used as a source image. Possible changes in the imaging geometry should be taken into consideration.

An option for increasing the number of source projections could be implemented. In cases, where more projections are available, a more accurate model could be generated using all available data.

References

1. J. Lötjönen, I. E. Magnin, L. Reinhardt, J. Nenonen, and T. Katila, "Automatic Reconstruction of 3D Geometry Using 2D Projections and a Geometric Prior Model", *Lecture Notes in Computer Science 1679: Medical Image Computing and Computer-Assisted Intervention, MICCAI99,* C. Taylor, A. Colchester (Eds.), Springer, pp. 192-201, 1999.
2. J. Lötjönen, P-J. Reissman, I.E. Magnin and T. Katila, "Model Extraction from Magnetic Resonance Volume Data Using the Deformable Pyramid", *Medical Image Analysis* **3**, 387-406, 1999.
3. J. Lötjönen, P-J. Reissman, I. E. Magnin, J. Nenonen and T. Katila, "A Triangulation Method of an Arbitrary Point Set for Biomagnetic Problems", *IEEE Trans. Magn*. **34**, 2228-2233, 1998.
4. M. Valkonen, *Combining fluroscopic imaging information with Body Surface Potential Mapping measurements*, Master's Thesis, Helsinki University of Technology, Finland, 2000.
5. O. Gerard and S. Makram-Ebeid, "Automatic Contour Detection by Encoding Knowledge into Active Contour Models", *Proceedings Fourth IEEE Workshop on Applications of Computer Vision (WACV'98)*, pp. 115-120, 1998.
6. J. Liang, T. McInerney, and D. Terzopoulos, "Interactive Medical Image Segmentation with United Snakes", *Lecture Notes in Computer Science 1679: Medical Image Computing and Computer-Assisted Intervention, MICCAI99,* C. Taylor, A. Colchester (Eds.), Springer, pp. 116-127, 1999.